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(54) Title: DYNAMIC RANGE EXTENSION TECHNIQUES FOR A WAVEFRONT SENSOR INCLUDING USE IN OPHTHALMIC MEASUREMENT

(57) Abstract: An ophthalmic error measurement system includes a projecting optical system delivering light onto a retina of an eye, a pre-correction system which compensates a light beam to be injected into the eye for aberrations in the eye, the pre-correction system being positioned in between the projecting optical system and the eye, and imaging system which collects light scattered by the retina, and a detector receiving light returned by the retina from the imaging system. Use of the pre-correction system allows the end-to-end aberrations of the ocular system to be analyzed. The use of a pre-correction system also allows use of a minimized spot size on the retina, and all of its attendant advantages.

DYNAMIC RANGE EXTENSION TECHNIQUES FOR A WAVEFRONT SENSOR INCLUDING USE IN OPHTHALMIC MEASUREMENT

5

Field of the Invention

The present invention is directed to measurement of the refractive error in the eye, more particularly to methods and techniques for compiling a topographic mapping of these refractive errors.

10

BACKGROUND OF THE INVENTION

Measurements of aberrations in an eye are important for diagnosis of visual defects and assessment of acuity. These measurements and their accuracy become increasingly important in light of the growing number of ways, both surgical and non-surgical, that 15 aberrations can be corrected. These corrections rely on accurate, precise measurements of the entire ocular system, allowing successful screening, treatment and follow-up. Enhancements in the accuracy of ocular measurements may aid in improving the identification of patients in need of correction and the performance of the correction itself.

There are a number of current methods used to measure performance of the ocular 20 system. The most widely used and well established are psycho-physical methods, i.e., methods relying on subjective patient feedback. The oldest of the psycho-physical methods is the foreopter or trial lens method, which relies on trial and error to determine the required correction. There are psycho-physical methods for measuring visual acuity, ocular modulation transfer function, contrast sensitivity and other parameters of interest.

In addition to these subjective methods, there are also objective methods for assessing 25 the performance of the ocular system. Such objective methods include corneal topography, wavefront aberrometry, corneal interferometry, and auto-refraction. Many of these methods

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only measure the contribution of specific elements to the total refractive error. For example, much work has been directed to measuring the topography of the cornea and characterizing the corneal layer. However, the corneal shape only contributes about 30-40% of the total refractive error in most cases. In order to measure the bulk of the refractive error and to 5 provide a complete mapping for diagnosis and correction, additional information and measurements are needed.

Another method for determining the refraction of the eye is auto-refraction, which uses a variety of techniques to automatically determine the required corrective prescription. These automated techniques include projecting one or more spots or patterns onto the retina, 10 automatically adjusting optical elements in the auto-refractor until the desired response is achieved, and determining the required correction from this adjustment. However, auto-refractors are not considered especially reliable. Further, auto-refractors measure only lower order components of the aberrations, e.g., focus and astigmatic errors.

Recently, the eye has started being considered as an optical system, leading to the 15 application of methods previously used for other optical systems to the measurement of the eye. These methods include interferometry and Shack-Hartmann wavefront sensing. These techniques are of particular interest because they measure the complete aberrations of the eye. This additional information allows measurement of non-uniform, asymmetric errors that may be affecting vision. Further, this information may be linked with any of the various 20 corrective techniques to provide improved vision. For example, U. S. Patent No. 5,777,719 to Williams describes the application of Shack-Hartmann wavefront sensing and adaptive optics for correcting ocular aberrations to make a super-resolution retina-scope. U.S. Patent No. 5,949,521 to Williams et al. describes using this information to make better contacts, intra-ocular lenses and other optical elements

25 Wavefront aberrometry measures the full, end-to-end aberrations through the entire optics of the eye. In these measurements, a spot is projected onto the retina, and the resulting returned light is measured with an optical system, thus obtaining a full, integrated, line-of-

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sight measurement of the eye's aberrations. A key limitation of the instruments used in these measurements is the total resolution, which is ultimately limited by the lenslet array of the instrument. However, selection of the lenslet array is itself limited by several factors, most importantly the size of the spot projected onto the retina.

5 A schematic illustration of the basic elements of a two dimensional embodiment of a Shack-Hartmann wavefront sensor is shown in Figure 2. A portion of an incoming wavefront 110 from the retina is incident on a two-dimensional lenslet array 112. The lenslet array 112 dissects the incoming wavefront 110 into a number of small samples. The smaller the lenslet, the higher the spatial resolution of the sensor. However, the spot size from small
10 the lenslet, due to diffraction effects, limits the focal length that may be used, which in turn leads to lower sensitivity. Thus, these two parameters must be balanced in accordance with desired measurement performance.

15 Mathematically, the image on the detector plane 114 consists of a pattern of focal spots 116 with regular spacing d created with lenslets 112 of focal length f , as shown in Figures 3. These spots must be distinct and separate, i.e., they must be readily identifiable. Thus, the spot size ρ cannot exceed $\frac{1}{2}$ of the separation of the spots. The spot separation parameter N_{FR} can be used to characterize the lenslet array 12 and is given by:

$$N_{FR} = \frac{d}{\rho} \quad (1)$$

20 The relationship between the size of a lens and the focal spot it creates, where λ is the wavelength of the light, is given by:

$$\rho = 1.22 \frac{f\lambda}{d} \quad (2)$$

for a round lens or

$$\rho = \frac{f\lambda}{d} \quad (3)$$

for a square lens. Thus, for a square lens, the separation parameter can be given by:

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$$N_{FR} = \frac{d^2}{f\lambda} \quad (4)$$

This is also known as the Fresnel number of the lenslet. To avoid overlapping focal spots, $N_{FR} > 2$. In practice, the Fresnel number must be somewhat greater than two to allow for a certain dynamic range of the instrument. The dynamic range of a Shack-Hartmann 5 wavefront sensor can be defined as the limiting travel of the focal spot such that the edge of the spot just touches the projected lenslet boundary, given by:

$$\theta_{\max} = \frac{\frac{d}{2} - \rho}{f} \text{ or} \quad (5)$$

$$\theta_{\max} = \frac{d}{2f} - \frac{\lambda}{d} = \left[\frac{N_{FR}}{2} - 1 \right] \frac{\lambda}{d} \quad (6)$$

Thus, the dynamic range is directly proportional to the separation parameter and the lenslet 10 size.

A particularly useful arrangement for a Shack-Hartmann wavefront sensor ocular measuring system places the lenslet array in an image relay optical system at a plane conjugate to the pupil or corneal surface. In this configuration, the spot size on the detector of the wavefront sensor is given by:

$$\rho_2 = \frac{1}{M} \frac{f_L}{f_e} \rho_1 \quad (7)$$

where M is the magnification of the imaging optics, f_L is the focal length of the lenslet array, f_e is the focal length of the eye and ρ_1 is the spot size on the retina.

Comparing Equations (5) and (7), it is evident that the dynamic range of the wavefront sensor is limited by the size of the spot ρ_1 projected on the retina. For a practical system, the 20 dynamic range must be able to resolve errors in the optical systems. Thus, the dynamic range is a key limited parameter of the entire system design. In previous implementations of the Shack-Hartmann wavefront sensor used for ocular measurement, the dynamic range has been increased by increasing the size of each lenslet. However, the eye itself can have significant

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aberrations. Thus, any beam projected into the eye will become aberrated, spreading the focal spot and increasing the spot size ρ_1 on the retina.

Various techniques have been implemented to address this problem. A small diameter beam has been used so that the total wavefront error is minimized across the injected beam.

5 Another proposed solution projects the light into the eye at the focal point of a long focal length lens, operating as a field lens so that the size of the focal spot is not affected by the eye aberrations. In practice, for both of these cases, the beam is still somewhat large and is increased in size by the aberrations of the ocular system.

Another limitation on the dynamic range of the system is the sampling size. With a
10 large spot on the retina, the sample size of the wavefront sensor must be increased to allow even a minimal dynamic range to be realized. For ocular systems with strong aberrations, such as found in people with large astigmatism or for those having undergone LASIK, the aberrations over each lenslet are sufficient to degrade the lenslet focal spot. Thus, the system is limited not just by focal spot overlap, but by the fact that the focal spots themselves fade
15 out or are difficult to track. Using a small sample size does not allow sufficient light to be gathered, since the light is scattered by the retina into a large number of focal spots. Due to safety considerations, the input power may not be increased to compensate for this scattering.

SUMMARY OF THE INVENTION

20 The present invention is therefore directed to measurement of refractive errors of an eye that substantially overcomes one or more of the problems due to the limitations and disadvantages of measurements of the related art.

It is an object of the present invention to measure the end-to-end aberrations of the eye with sufficient accuracy and dynamic range in a practical manner.

25 It is a further object of the present invention to project a light beam into an ocular system so as to minimize the size of the focal spot on the retina.

It is another object of the present invention to use this smaller focal spot to allow

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much greater sampling density of the ocular system, thereby enhancing the accuracy and dynamic range.

It is yet another object of the present invention to make a practical, low cost system, available for use in a clinical setting.

5 At least one of the above and other objects may be realized by providing a system for measuring errors in an eye including a projecting optical system which delivers light onto a retina of the eye, a pre-correction system which compensates a light beam to be injected into the eye for aberrations in the eye, the pre-correction system being positioned in between the projecting optical system and the eye, an imaging system which collect light scattered by the
10 retina, and a detector receiving light from the retina collected by the imaging system.

The detector may be a Shack-Hartmann wavefront sensor, a shearing interferometer, a Moiré deflectometer, or other passive phase measurement systems. The pre-correction system may include a telescope having at least one movable lens, fixed lenses inserted at an intermediate image plane, adaptive optical elements, and/or a cylindrical telescope. The pre-
15 correction system may correct for focus and/or astigmatism errors in the eye. The telescope may be arranged so that a fixed lens of the telescope is one focal length away from the eye. Components used in the pre-correction system may also be used in the imaging system.

The pre-correction system may include a feedback loop which determines an appropriate pre-correction to be supplied by the pre-correction system. The feedback loop
20 may include a detector receiving light returned from the retina, a processor comparing detected light with a desired feature of the light and adjusting at least one parameter of the pre-correction system in accordance with the comparison. The feedback loop may further include a return optical system for gathering the light from the retina. The return optical system may include the pre-correction system. The desired feature may be a minimized spot
25 size on the retina.

The system may include an aperture that limits the angular dynamic range of the system. The system may further include a polarizing beam splitter between the eye and the

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wavefront sensor. The system may include an aligner that determines an appropriate eye alignment of the system. The projecting optical system may provide light to the eye at an angle to a central axis of the eye. The system may include an additional optical system between the detector and the eye. The system may include a power monitor which monitors 5 power of the light beam being injected in the eye. The system may include an eye position detection system including a target projected on the eye, a position detector sensing the eye, and an adjustment system which adjusts a position of the system relative to the eye until the eye is in focus on the detector.

These and other objects of the present invention will become more readily apparent 10 from the detailed description given hereinafter. However, it should be understood that the detailed description and specific examples, while indicating the preferred embodiments of the invention, are given by way of illustration only, since various changes and modifications within the spirit and scope of the invention will become apparent to those skilled in the art from this detailed description.

15

BRIEF DESCRIPTION OF THE DRAWINGS

The foregoing and other objects, aspects and advantages will be described with reference to the drawings, in which:

Figure 1 is a schematic top view of the measurement system of the present invention;

20 Figure 2 is a schematic side view of the basic components of a Shack-Hartmann wavefront sensor;

Figure 3 schematically illustrates the relationship between the size of the lens, its focal length and the spot size;

Figures 4A-4C schematically illustrate the spot size for different configurations;

25 Figures 5A-5B schematically illustrate off-axis injection of the light into the eye and the blocking of the reflected light from entering the wavefront sensor;

Figure 6 is a schematic illustration of a configuration of the present invention using a

fixed telescope and an adjustable telescope;

Figure 7 is a schematic illustration of a configuration of the present invention using a variable lens;

5 Figure 8 is a schematic illustration of a cylindrical telescope for use with the present invention; and

Figure 9 is a schematic illustration of a configuration of the present invention using a corrective lens.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

10 As noted above, the key to designing a practical ocular wavefront sensor system is how the light is injected into the eye. Since ocular refractive errors can be large, e.g., up to 20 diopters, the degradation of the injected beam can be significant. Further, it is difficult to design a wavefront sensor that has sufficient range to directly measure an extremely large refractive error. In accordance with the present invention, the spot projected on the ocular system is predistorted in a manner that compensates for the eye's fundamental aberrations.

15 This allows the spot returned to the wavefront sensor to be well formed and minimally affected by the refractive errors. The small size of the spot allows small lenslets to be used while maintaining sufficient dynamic range to measure even large, high order aberrations. Since the light is tightly focused on the retina, the light is only scattered from a small region.

20 When this small region is imaged onto the focal plane of the wavefront sensor, the light is concentrated onto a small group of pixels. Thus, even though the reflected light must be divided among a larger number of lenslets, each focal spot is brighter than in the conventional methods. Further, the greater sampling density leads to smaller wavefront aberrations across the aperture of each lenslet.

25 A system for such error measurement employing pre-compensation is shown in Figure 1. The ocular wavefront measurement system shown therein generally includes a projection system for projecting light into the eye, a system for pre-correcting the injected light for

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ocular aberrations, a system for collecting light, a system for determining the pre-correction, and a system for measuring the collected light.

The projection system shown in Figure 1 includes a light source 12, e.g., a laser, a laser diode, LED, or a super-luminescent diode, supplied to an optical fiber 14. For safety 5 reasons, the light source is preferably a pulsed light source, is limited to a small power, is outside the normal visual detection range, e.g. infrared, and/or is directly collimated with an appropriate lens. The optical fiber may be a polarization maintaining fiber. The light leaving the optical fiber 14 is provided to a collimating lens 16. The use of an optical fiber 14 to deliver light from the light source 12 simplifies the collimating lens 16, since the fiber exit 10 mode acts as a diffraction-limited point source. The collimating lens 16 is preferably rigidly mounted to the fiber 14. The collimated beam is then truncated to a desired size by an aperture 18. If needed, a polarizer 20 may be provided for polarizing the collimated beam. A polarizing beam splitter 22 directs the light from the projection system to the rest of the ocular measuring system.

15 Alternatively, the light source 12 may be provided alone, i.e., without the use of the fiber 14. The light from the light source 12 itself is then collimated by a collimating lens. While light sources used for ophthalmic measurement typically have a high degree of 20 astigmatism, by using only a portion of the beam, e.g., 10- 25 %, typically from the center of the beam, the wavefront error over the beam is small enough that the beam size is substantially stable over the distance traversed in the ophthalmic measurement system. In other words, even though the beam is still astigmatic, the beam shape does not change while traversing the ophthalmic measurement system due to this astigmatism, so the astigmatism does not influence the measurement. The light may be polarized as required.

25 The light from the projection system is reflected by the polarizing beam splitter 22 and directed to a pre-compensation system, shown in Figure 1 as a telescope 30. The telescope 30 includes lenses 32, 34 with an aperture 36 in between. The telescope 30 may be adjusted by moving the lenses relative to one another. This adjustment is to provide the desired pre-

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correction for the injected beam by adding defocus that just compensates for the spherical equivalent defocus of the ocular system being measured. The light from the telescope is directed by a beam splitter 38 to an ocular system 40 under measurement. The injected beam is focused by the ocular system 40 to a focal spot 42 on the retina of the ocular system 40.

5 Light from this focal spot 42 is scattered or reflected by the retina.

The returned light is collected by the cornea and lens of the ocular system 40 and is approximately collimated. The beam splitter 38 directs the beam from the ocular system back to the telescope 30. The same position of the lenses 32, 34 of the telescope 30 corrects for the defocus aberrations of the ocular system 40 so that light arrives at a wavefront sensor

10 50 collimated to within the dynamic range of the sensor. The aperture 36 blocks any rays outside the angular dynamic range of the wavefront sensor 50 so that no mixing or measurement confusion occurs. When the wavefront sensor 50 is a Shack-Hartmann sensor, the focal spots cannot collide, interfere or cause confusion with adjacent focal spots. The light from the telescope now passes through the polarizing beam splitter 22, since the

15 interaction with the retina will rotate the polarization of the light from the input polarization. The wavefront sensor 50 may be a Shack-Hartmann wavefront sensor, a shearing interferometer, a Moiré deflectometer or any other passive phase measurement sensor. When the wavefront sensor 50 is a Shack-Hartmann wavefront sensor, the wavefront sensor 50 includes the elements shown in Figure 2.

20 The proper position of the lenses 32, 34 of the telescope 30 may be determined in a number of ways. In a preferred embodiment, an additional sensor 60 is used with a beam splitter 62 and a focusing lens 64 to create an image of the light incident upon the retina. The proper position of the lenses 32, 34 in the telescope 30 is determined by minimizing the spot size 42 on the back of the retina, performed by comparing the spot sizes from different 25 positions of the lenses 32, 34 in the telescope 30. If the ocular system 40 is arranged to be one focal length of the objective lens 34 away from the lens 34, then the telescope 30 will be insensitive to changes in magnification or other errors. The wavefront sensor 50 should be

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arranged to be at the conjugate image plane to the ocular system 40. Preferably, the wavefront sensor 50, the retinal imaging sensor 60, the projection optics 16, 18, 20, the polarizing beam splitter 22, the beam splitter 62, and the focusing lens 64 are mounted on a platform 70 which is mounted in a moving stage 72. This allows the relative position of the telescope lenses 32, 34 to be varied while fixing the position of the remaining elements on the platform 70. The use of the optical fiber 14 allows the light source to be mounted off the platform 70, minimizing the mass of the elements moved by the translation stage 72. A processor 68 may be included to control movement of the translation stage 72 and to allow data processing, analysis and/or display.

As an additional safety measure, a small portion of the beam incident on beam splitter 38 is transmitted to a lens 44 which focuses the light onto a power monitor 46. The output of this power monitor 46 may be used to shut down the system if the power exceeds the safety limits of the system or to alter the power supplied to the light source 12 to reduce the power output by the light source in a known manner.

To measure the proper eye position relative to the measuring system, an additional detector 80 is included. Imaging optics 82 are designed such that the iris or cornea will be in focus for only a narrow region of space. A mirror 84 may be used to direct light onto the iris detector 80. The position of the system relative to the eye is adjusted until the iris or cornea is detected. The detection may be indicated to a user on an indicator 86. Preferably, this detection is used just during patient alignment and only uses a small percentage, e.g., less than 10% of the light.

To insure that the patient is viewing the correct line of sight, a target 90 is made visible through a beam splitter 94. The target 90 is imaged at infinity through a lens 92. The target position may be varied by moving the target relative to the lens 92 to present targets that are either in focus or slightly out-of-focus to minimize patient accommodation. Movement of the target 90 closer to the lens 92 stimulates near vision accommodation, allowing measurement of near vision visual acuity or the target may be arranged with the

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image past infinity to measure distance vision. The patient merely attempts to focus on the target. A light source behind the target is electronically controlled to adjust the target brightness and the position of the target is also electronically adjustable.

Thus, the telescope 30 is used to pre-compensate the injected light and to compensate for the returned wavefront to minimize the total wavefront error incident on the wavefront sensor. In the related art, telescopes have been used to relay image the light onto the wavefront sensor and to compensate for strong spherical and cylindrical aberrations, but the light was injected separately. This separate handling is due to strong back reflections that occur even for lenses having anti-reflection coatings thereon. Since the returned light from the retina may be very weak, even a small reflection from the lenses can quickly dominate the measurement and saturate the wavefront sensor 50. There are several ways of dealing with the problem. First, as shown in Figure 1, polarized light and a polarizing beam splitter in conjunction with a quarter-wave plate may be used. Off axis parabolas or other curved mirrors may be used to direct the light to the telescope. The light may be injected off axis, so that any reflected light from the cornea is filtered out by the apertures of the system, as shown in Figures 5A and 5B. Figure 5B illustrates how the light reflected by the cornea of the eye 40 is blocked by the aperture 36 from entering the wavefront sensor and influencing the measurement. The use of one or more of these schemes is sufficient to allow pre-compensation of the injected beam in accordance with the present invention without introducing unwanted reflections.

As an alternative, a second telescope may be used in conjunction with the first telescope to increase the dynamic range by providing an alternative location for the filtering aperture. Thus, one telescope can be completely fixed, while the other has a degree of freedom allowing movement until the lenses of the two telescopes are in contact. Such a configuration is shown in Figure 6, in which a fixed telescope 51 with lenses 52, 54 and aperture 56, is used to supply light to the wavefront sensor 50. This is in conjunction with the elements discussed above regarding Figure 1. For simplicity, only the essential elements

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of the light delivery system 14; the collimating lens 16, the polarizing beam splitter 22, the adjustable telescope 30, and the eye 40, have been shown.

Compensation of astigmatism of the ocular system and of the injected beam may be achieved in the following ways. The telescope 30 may be a cylindrical lens telescope or a 5 pair of positive and negative lenses. Such a cylindrical lens configuration is shown in Figure 8, in which a pair of cylindrical lenses 132, 134 is used in place of lenses 32, 34. The spacing s between the lenses may be adjusted to increase or decrease power of the telescope. The angle of the pair 120, 122 is adjusted relative to the axis of the transmission path. This 10 complicates the instrument, but provides for a better beam projected into the eye, requiring a wavefront sensor of only limited dynamic range, since both spherical and cylindrical aberrations would be subtracted from the wavefront, and only higher order terms would remain.

Alternatively, a high dynamic range wavefront sensor can be used. Since, in accordance with the present invention, only a small beam is injected into the eye, which will 15 only pick up only a small wavefront aberration across its aperture, the focal spot on the eye will still be quite small, even with some astigmatism. Thus, cylindrical compensation is usually not needed. While some distortion will take place, it will be limited in size and an adequately small spot will still be realized. A high dynamic range wavefront sensor corresponds to the use of a smaller focal length for the wavefront sensor lenslet array, as set 20 forth in Equations (3) and (7). While the use of only spherical lenses will result in a loss of accuracy, the larger number of measurements afforded by the smaller lenslet array will sufficiently compensate for this degradation.

An alternative to using the telescope with a movable lens, as shown in Figure 1, for 25 correcting base aberrations of the eye in the injected and reflected wavefront includes placing a corrective lens in front of the eye. If this lens is not a contact lens, it cannot be placed at the actual pupil plane of the eye, as shown in Figure 9, in which a corrective lens 35 is placed adjacent to the eye 40. Thus, there will always be some magnification introduced by the

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combination of the refractive error of the eye and the correcting lens. Since it is difficult to set or know the vertex distance of the corrective lens, this magnification would be poorly known at best, and introduce error into the entire measurement.

Another alternative includes using fixed or variable lenses. Ideally, these lenses are placed at an optical plane that is conjugate to the surface of the eye. Since it is also desirable for the wavefront sensor to be at this plane, a second telescope will need to be used in series with the first telescope. Further, since all of the lenses are fixed, some means will be needed for changing the various pre-corrector lenses in a known manner to achieve the proper result. A lens 37 in Figure 7 may be from a trial lens kit, such as is commonly used for measuring a patient's manifest refraction, but is limited to the prescription accuracy. Alternatively, the lens 37 in Figure 7 may be a variable focal length lens, e.g., adaptive optics, liquid crystal displays, deformable mirrors. The focal lengths of these elements may be controlled electronically, e.g., by the processor 68 shown in Figure 1, rather than by movement. Either of these configurations is shown in Figure 7, in which the lens 37 may be a trial lens or a variable focal length lens. The applicability of these configurations and the telescope configuration is shown in Figure 4A-4C, in which the size of the spot in a myopic eye alone is shown in Figure 4A, the size of the spot size with correction with a lens 37 is shown in Figure 4B and the spot size with the adjustable telescope 30 is shown in Figure 4C. As can be seen, both configurations in Figures 4B and 4C result in the desired small spot size of the present invention.

By pre-compensating for aberrations of the eye in the injected beam in accordance with the present invention, a small focal spot can be created on the retina. This small focal spot will concentrate light more, allowing the light to be divided into a larger number of focal spots. This, in turn, allows higher spatial resolution and the use of lower injected light power. Higher spatial resolution means that the assumption that each lenslet measures only tilt is valid over a much larger range. Higher spatial resolution also leads to greater dynamic range and accuracy. Higher dynamic range means that measurement of even high order terms

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of aberration can be accomplished accurately, without significant degradation of the measurement.

While the present invention is described herein with reference to illustrative embodiments for particular applications, it should be understood that the present invention is not limited thereto. Those having ordinary skill in the art and access to the teachings provided herein will recognize additional modifications, applications, and embodiments within the scope thereof and additional fields in which the invention would be of significant utility without undue experimentation. Thus, the scope of the invention should be determined by the appended claims and their legal equivalents, rather than by the examples given.

What is Claimed:

1. A system for measuring errors in an eye comprising:
 - a projecting optical system which delivers light onto a retina of the eye;
 - a pre-correction system which compensates a light beam to be injected into the eye for aberrations in the eye, the pre-correction system being positioned in between the projecting optical system and the eye;
 - an imaging system which collects light scattered by the retina; and
 - a detector receiving light returned by the retina from the imaging system.
- 10 2. The system of claim 1, wherein the detector is a Shack-Hartmann wavefront sensor.
3. The system of claim 1, wherein the detector is a shearing interferometer.
- 15 4. The system of claim 1, wherein the detector is a Moiré deflectometer.
5. The system of claim 1, wherein the pre-correction system comprises a telescope having at least one movable lens.
- 20 6. The system of claim 1, wherein the pre-correction system comprises fixed lenses inserted at an intermediate image plane.
7. The system of claim 1, wherein the pre-correction system comprises adaptive optical elements.
- 25 8. The system of claim 1, wherein the pre-correction system corrects for focus errors in the eye.

9. The system of claim 1, wherein the pre-correction system corrects for focus and astigmatism.
- 5 10. The system of claim 1, further comprising an aperture that limits the angular dynamic range of the system.
11. The system of claim 1, further comprising a polarizing beam splitter between the eye and the wavefront sensor.
- 10 12. The system of claim 1, further comprising an aligner that determines an appropriate eye alignment of the system.
- 15 13. The system of claim 1, wherein the projecting optical system provides light to the eye at an angle to a central axis of the eye.
14. The system of claim 1, further comprising an additional optical system between the detector and the eye.
- 20 15. The system of claim 1, further comprising a power monitor which monitors power of the light beam being injected in the eye.
- 25 16. The system of claim 1, further comprising an eye position detection system including a target projected on the eye, a position detector sensing the eye, and an adjustment system which adjusts a position of the system relative to the eye until the eye is in focus on the detector.

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17. The system of claim 1, wherein the pre-correction system is a cylindrical telescope.

18. The system of claims 1, wherein components used in the pre-correction system
5 also are used in the imaging system.

19. The system of claim 1, wherein the pre-correction system includes a feedback loop
which determines an appropriate pre-correction to be supplied by the pre-correction system.

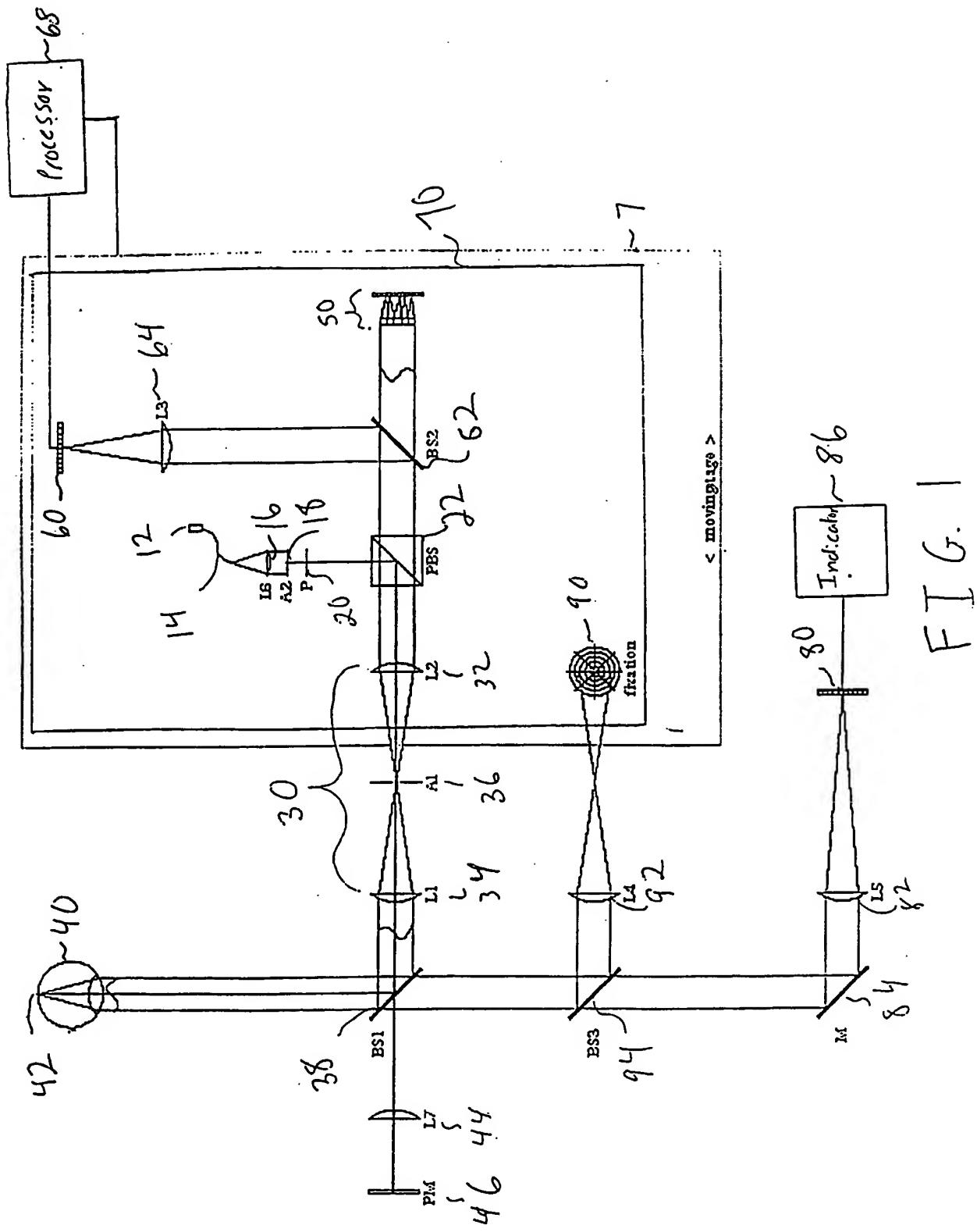
10 20. The system of claim 19, wherein said feedback loop includes a detector receiving
light returned from the retina, a processor comparing detected light with a desired feature of
the light and adjusting at least one parameter of the pre-correction system in accordance with
the comparison.

15 21. The system of claim 20, wherein said feedback loop further includes a return
optical system for gathering the light from the retina.

22. The system of claim 21, wherein said return optical system includes said pre-
correction system.

20 23. The system of claim 20, wherein said desired feature is a minimized spot size on
the retina.

25 24. The system of claim 5, wherein said telescope is arranged so that a fixed lens of
the telescope is one focal length away from the eye.



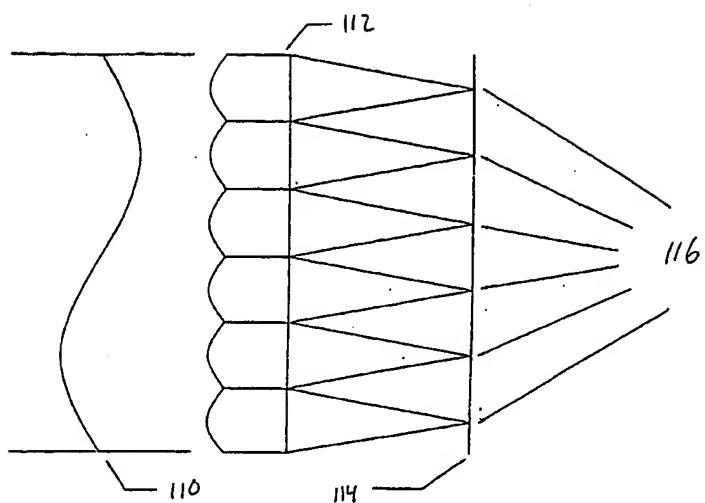


FIG. 2

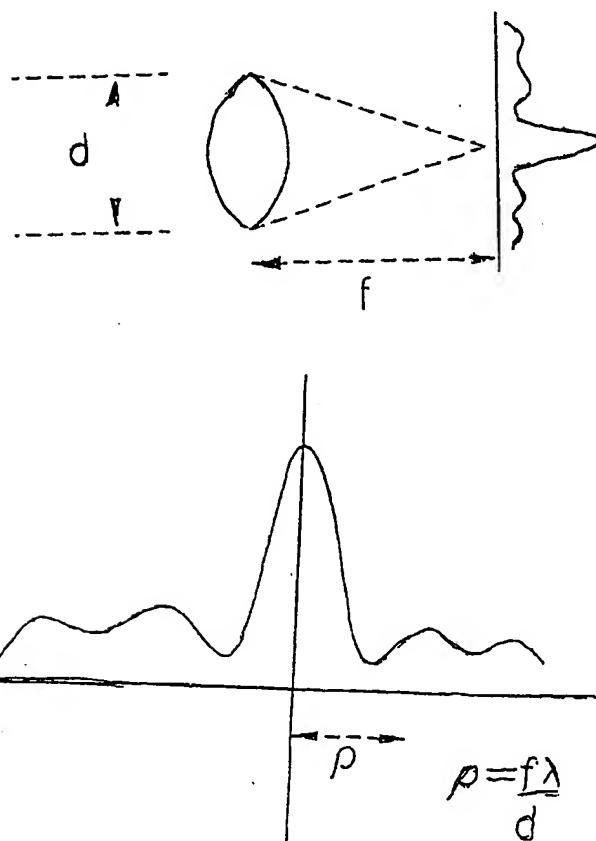


FIG. 3

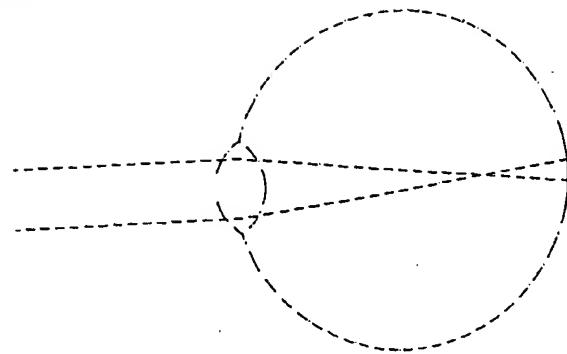


FIG. 4A

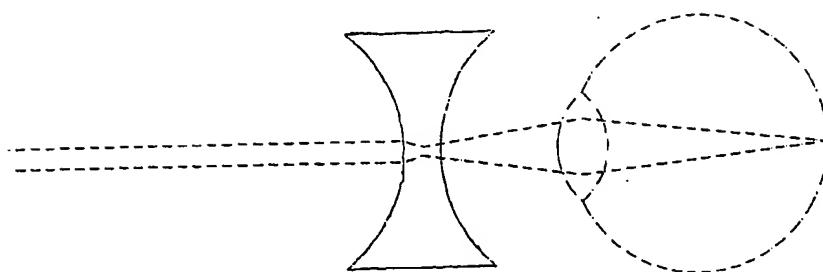


FIG. 4B

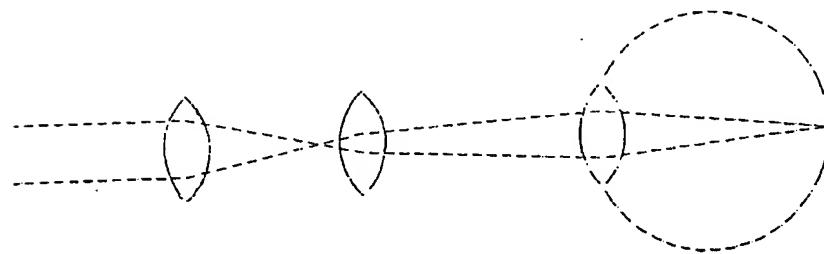


FIG. 4C

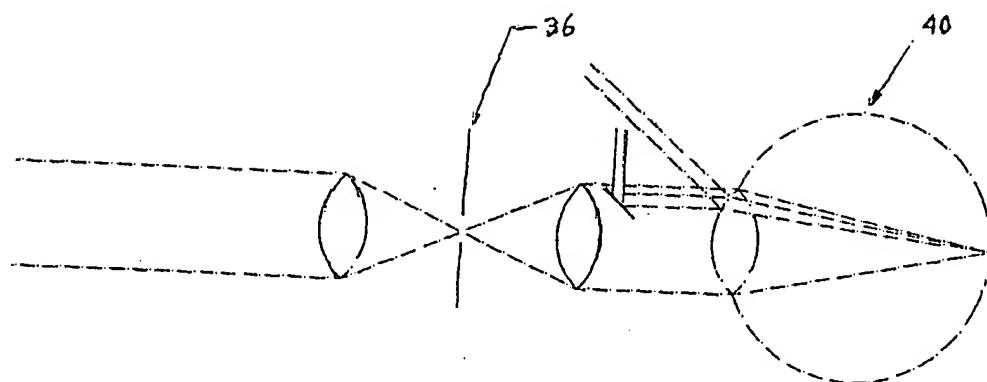


FIG. 5A

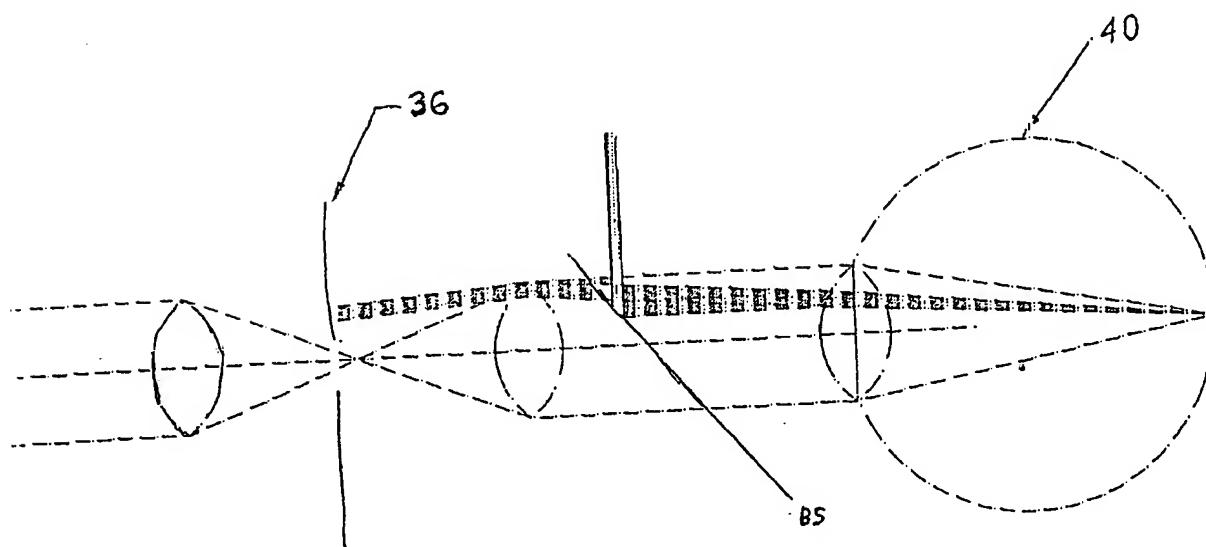


FIG. 5B

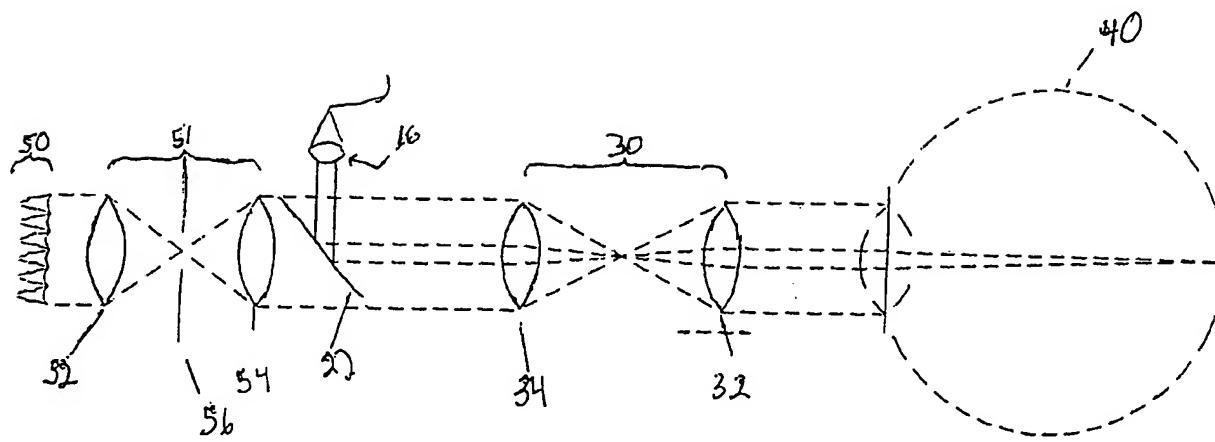


FIG. 6

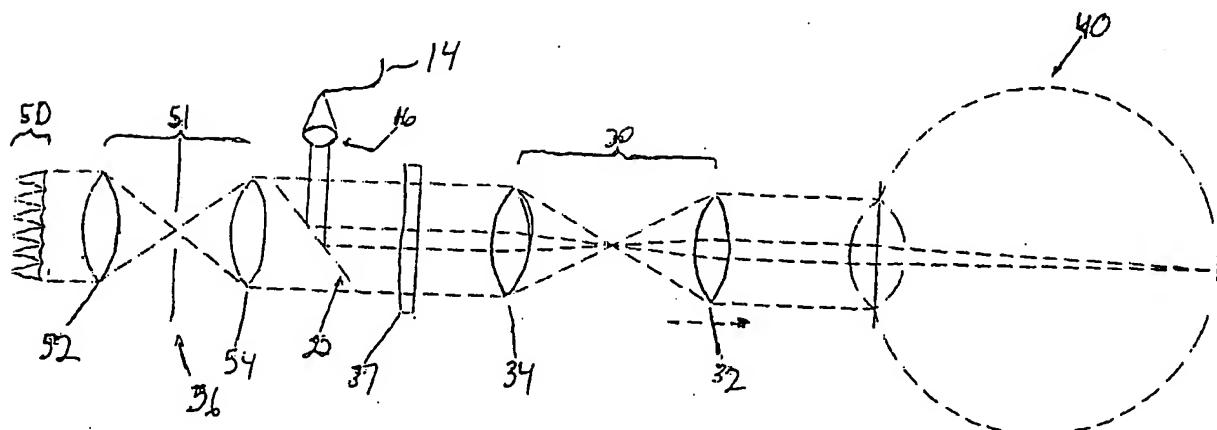


FIG. 7

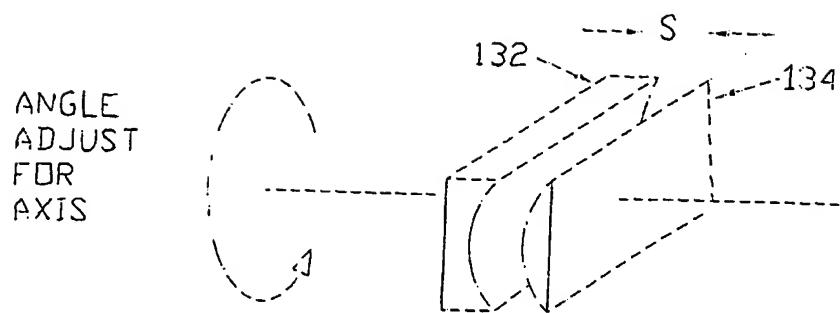


FIG. 8

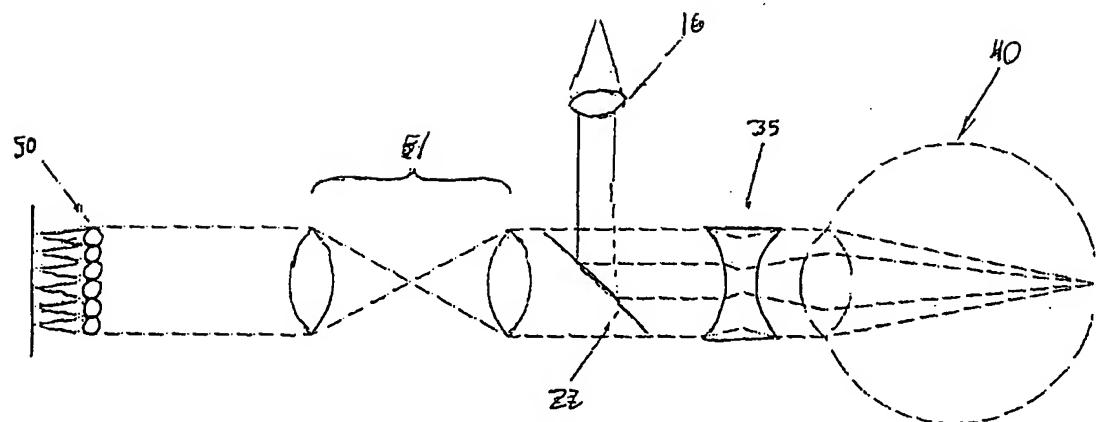


FIG. 9

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(71) Applicant (for all designated States except US): WAVEFRONT SCIENCES, INC. [US/US]; 14810 Central Avenue, SE, Albuquerque, NM 87123 (US).

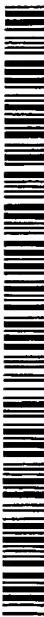
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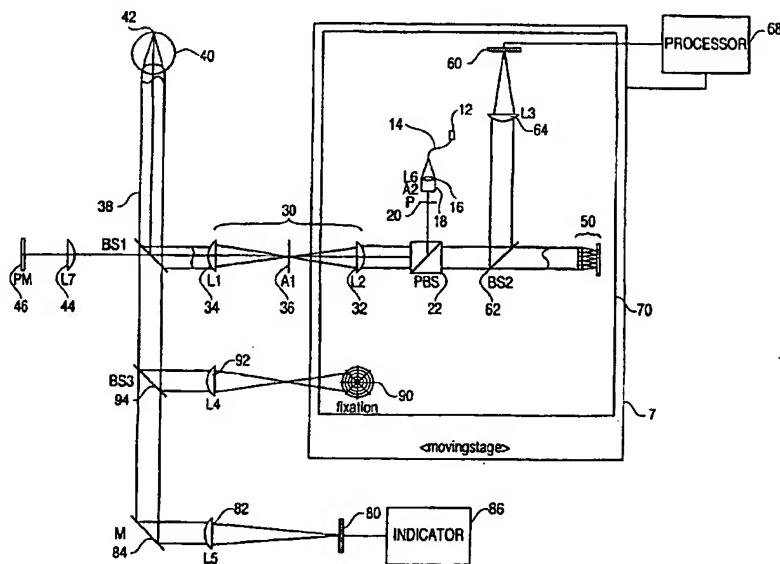
(75) Inventors/Applicants (for US only): NEAL, Daniel, R. [US/US]; 17 Eric Alan Lane, Tijeras, NM 87123 (US). ARMSTRONG, Darrell, J. [US/US]; 910 Santa Anna Avenue S.E., Albuquerque, NM 87123 (US). GRUETZNER,

[Continued on next page]

(54) Title: DYNAMIC RANGE EXTENSION TECHNIQUES FOR A WAVEFRONT SENSOR



WO 01/058339 A3



(57) Abstract: An ophthalmic error measurement system includes a projecting optical system delivering light onto a retina of an eye, a pre-correction system which compensates a light beam to be injected into the eye for aberrations in the eye, the pre-correction system being positioned in between the projecting optical system and the eye, and imaging system which collects light scattered by the retina, and a detector receiving light returned by the retina from the imaging system. Use of the pre-correction system allows the end-to-end aberrations of the ocular system to be analyzed. The use of a pre-correction system also allows use of a minimized spot size on the retina, and all of its attendant advantages.



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Minimum documentation searched (classification system followed by classification symbols)
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Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

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Y	column 4, line 26 -column 5, line 15; figures 1-7	2,5
A	---	6-24
Y	DE 42 22 395 A (AMTECH GESELLSCHAFT FÜR ANGEWANDTE MICROCOMPUTER TECHNIK MBH) 13 January 1994 (1994-01-13)	2
A	column 4, line 56 -column 5, line 17; figures 1-3	6-24
Y	US 5 258 791 A (PENNEY ET AL.) 2 November 1993 (1993-11-02)	5
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